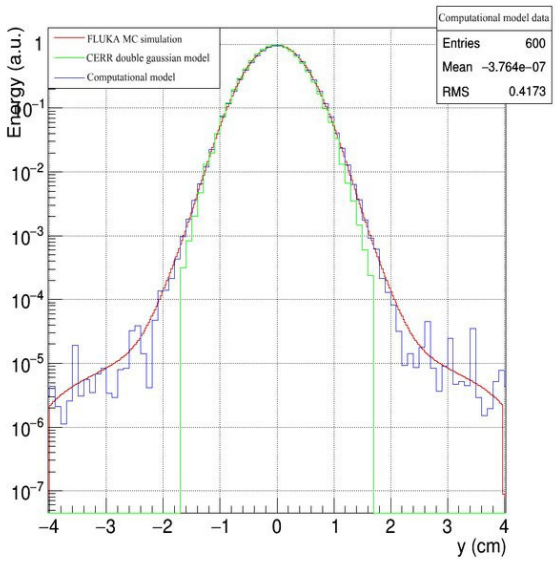

Fig.1 Comparison between Fluka MC simulation (red line), dose distribution evaluated with CERR using Double Gaussian model (green line) and dose predicted by standalone computational model (blue line) for a proton initial beam energy of 60MeV, initial σ of 0.4cm, at a normalized depth 0.9 (assuming Bragg peak position at 1)

The preliminary results show that the model is able to predict dose with better accuracy than DGA.
So, this model could be a step forward in lateral dose prediction for current TPS and as a future application, a useful tool for online dose-evaluation and adaptive planning; next step will be the implementation of different media and geometries.

Keywords:
Proton-therapy, Treatment Planning System, Molière lateral dose distribution.

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A clinical protocol for Simultaneous Integrated Boost for proton treatment of Head and Neck carcinoma

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Purpose: To define the clinical protocol for Head and Neck (H&N) patients treated with Proton Pencil Beam Scanning (PBS) at Paul Scherrer Institute (PSI, Villigen CH) with Simultaneous Integrated Boost (SIB). The focus was mainly to define the optimization parameters and the normalization procedure to obtain a steep gradient between the elective and the boost volume, in order to reduce over-dosage in the fall off area of the boost.
Material and Methods: 7 patients, treated at PSI for different H&N carcinomas, were selected for this study. They were all originally treated with two sequential series at 2GyRBE; the 1st one up to 54GyRBE to the elective Planning Target Volume (PTV1), followed by a second series up to 72GyRBE for the boost target volume (PTV2). In this study, all have been re-planned using a SIB regimen of 1.8GyRBE (up to 54GyRBE) to PTV1 and 2.36GyRBE (up to 70.8GyRBE) to PTV2 for a total of 30 fractions. Dose constraints to the OARs were kept as for the sequential approach [1]. All plans were designed on the PSIPlan Treatment Planning System using Intensity Modulated Proton Therapy (IMPT) with non-coplanar fields. The prescription dose of 100% corresponded to the average dose to PTV1, while the dose to PTV2 was optimized prescribing a boosting factor based on the ratio of the two different prescribed dose levels (i.e. 54GyRBE and 70.8 GyRBE). Depending on the ratio of the PTV1 and PTV2 volumes the boosting factor was prescribed to PTV2 or to a smaller structure (GTV or PTV2-3mm). To guarantee dose homogeneity in the dose gradient region between PTV1 and PTV2, the optimized SIB plans were normalized such that 54GyRBE corresponded to the average dose to [PTV1-(Boost volume+3mm)]. This avoided over-dosages close to PTV2 being compensated by under-dosages at the PTV1 margin.
Results: The choice of the boosted volume, the boosting factor, the quality and conformity of the SIB plans were observed to be dependent on the volume ratio PTV2/PTV1. For the analyzed patients, this ranged from 28.8% to 61.7%. For differences in the PTV1 to PTV2 volume larger than 40%, either a PTV2-3mm or the GTV itself were selected for the boosting. Else no gradient could be observed. When compared to the nominal sequential plan, the SIB approach resulted in both a lower mean dose to the ring area (average value for SIB: 55.2±1.0 GyRBE; average value for sequential: 64.1±3.6 GyRBE), and 0.6±0.4% as average ring volume receiving the 95% of the dose prescribed to the boost (the same dose parameter was >10% for all the original plans). A similar sparing of OARs was obtained with both treatment schedules.
Conclusion: Planning H&N patients with SIB optimization resulted in dose distributions which guaranteed the PTV2 and PTV1 coverage and conformity whilst keeping dose to OARs within tolerance. Therefore this approach can be transferred to the clinical operation and has already been applied to a first patient.

Keywords: over-boosting, gradient, boosting factor

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Gamma Locator for Radionuclide Diagnostics Of Oncological Diseases

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Gamma locator is a handheld lightweight and compact gamma probe based on a scintillation crystal $LaBr_3$:Ce and silicone photomultiplier to be used for detection of gamma-radiation emitted by radionuclides such as ^{99m}Tc, ^{125}I, ^{111}In, ^{18}F [1].
There are two main applications of gamma locator: intraoperative detection of sentinel lymph nodes and non-invasive scanning the surface of the body. In the first case, a radiotracer is injected into the patient preoperatively and

surgeon checks the presence of metastasis in the lymph nodes after the removal of the tumor. In the second case, the gamma locator can detect tumors located superficially, and accurately determine their boundaries [2].
When selecting the scintillator the following requirements should be considered: high relative light yield compared to NaI(Tl), the high value of the effective atomic number. The lanthanum cerium bromine has demonstrated the best features: the light yield higher than that of NaI(Tl) (130%), high density, high atomic number provide high efficiency of photoelectric absorption of gamma rays, and the decay time of the order of tens of nanoseconds provides high temporal resolution of the detector. Silicon photomultiplier is a device for detection of low intensive and very fast (several hundred nanosecond duration) light flashes. SiPM is used due to its high detection efficiency, low bias voltage, compact dimensions and high gain of signals. Experimental studies have shown that a scintillator packaged together with the photodetector provides 4,1% FWHM energy resolution at 662 keV (Cs-137).
Gamma locator is constructed in cordless configuration and equipped with a lithium-ion battery. Indication is performed by an acoustic signal and LED. Adjusting of the bias voltage of the photodetector and the thresholds of the discriminator is carried out by changing the resistance of the trimmers.
The main technical characteristics of the prototype gamma locator were determined in the laboratory of NRNU MEPhI according to the NEMA NU3-2004 protocol [3]. Spatial resolution of gamma locator is the minimum distance between two point sources on which they can be resolved separately, or FWHM of the dependence of counting rate of the transverse distance between the detector and the source, and is measured to be 20 mm. Spatial selectivity is the polar angle, at which the count rate drops twice and it is measured to be 26 degrees. Sensitivity is 118 cps/MBq.
For the comparison performance of scintillation and semiconductor gamma locator the NEMA testing of the semiconductor CdTe-based commercial gamma probe was performed. The field of view of the CdTe gamma probe was formed with a lead collimator with 3 mm diameter aperture. Spatial resolution was measured to be 22 mm, and angular resolution is 30 degrees.

Keywords: gamma probe, miniature gamma detector, radioguided surgery

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Evaluation of the size of micrometric/nanometric dosimeters for use in radiotherapy and medical physics
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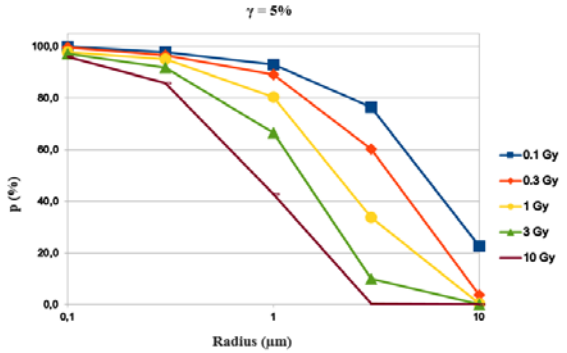
When treating tumors with radiotherapy, it is of utmost importance to ensure that the prescribed dose is accurately delivered to the target volumes. In that sense, *in-vivo* dosimetry in real time was recently implemented in radiotherapy departments. Dosimeter performance depends necessarily on physical and geometrical parameters (e.g. beam energy and distance from source to skin), which implies the use of correction factors. Implantable dosimeters are therefore preferable in order to minimize the need for corrections. They should be as small as possible, but still they should provide reliable measurements to comply with the requirements of clinical practice in routine radiotherapy. The state-of-the-art of these kind of dosimeters was the subject of a review elsewhere (1), which reported that implantable detectors of submillimetric size are currently available. The purpose of this study is to assess by Monte-Carlo simulations how much the size of such dosimeters can be decreased without jeopardizing their performance in a clinical environment.
First, the interaction of photons from a ^{60}Co source with water was simulated with a Monte-Carlo tool (2). The calculations were performed for 0.3, 0.1, 1, 3 and 10 Gy. Then, the distributions of specific energy were obtained for volumes representing dosimeters at nanometric and micrometric scales. Cylinders with equal radii of 0.3, 0.1, 1, 3 and 10 µm were used for this purpose. The mean specific energy $<z_t>$ was calculated for each case. To evaluate how the dosimeter size would impact its performance in a clinical scenario, the probability *p* that a dosimeter measurement falls outside a given interval defined around $<z_t>$ was estimated. Intervals were defined as $[<z_t>-\gamma<z_t> ; <z_t>+\gamma<z_t>]$ with γ equal to 3%, 5% and 10%.
The pattern of the distributions of specific energy evolves with dosimeter size and irradiation dose. Fixing the irradiation dose and decreasing the dosimeter radius or fixing the radius and decreasing the irradiation dose strongly widened the range in measured values of specific energy, but also increased the probability of yielding a non-null measurement. In turn, for higher doses and radii, distributions tend to Gaussian curves.



Concerning the probability of obtaining a measurement outside the defined interval, the larger the interval, the irradiation dose, and the dosimeter radius, the smaller this probability became (see figure above).
The simulation results showed that dosimeters at a nanometric scale are not able to yield statistically-reproducible measurements and are therefore unfit for use in clinical practice. Increasing the size to micrometric scale led to a decrease in the statistical fluctuations. Nevertheless, to have enough accuracy at routine clinical doses (approximately 2 Gy in the tumor volume), a dosimeter radius of at least 10 µm is required.

Keywords: radiotherapy; nano/microdosimeter; Monte-Carlo simulations

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From 2D to 3D: Proton radiography and proton CT in proton therapy: A simulation study